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Title NUCLEAR MEDICAL DIAGNOSTIC APPARATUS

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**APPLICATION ELEMENTS**

*See MPEP chapter 600 concerning utility patent application contents*

ADDRESS TO: Assistant Commissioner for Patents  
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Fee Transmittal Form (e.g. PTO/SB/17)  
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2. Specification Total Pages **33**
3. ☒ Drawing(s) (35 U.S.C. 113) Total Sheets **4**
4. ☒ Oath or Declaration Total Pages **2**
- a. ☒ Newly executed (original)
- b. ☐ Copy from a prior application (37 C.F.R. §1.63(d))  
(for continuation/divisional with box 15 completed)
- i. ☐ DELETION OF INVENTOR(S)  
Signed statement attached deleting inventor(s) named  
in the prior application, see 37 C.F.R. §1.63(d)(2) and  
1.33(b).
5. ☐ Incorporation By Reference (usable if box 4B is checked)  
The entire disclosure of the prior application, from which a copy of the  
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of the disclosure of the accompanying application and is hereby  
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**ACCOMPANYING APPLICATION PARTS**

6. ☒ Assignment Papers (cover sheet & document(s))
7. ☐ 37 C.F.R. §3.73(b) Statement (when there is an assignee) ☐ Power of Attorney
8. ☐ English Translation Document (if applicable)
9. ☐ Information Disclosure Statement (IDS)/PTO-1449 ☐ Copies of IDS Citations
10. ☐ Preliminary Amendment
11. ☐ White Advance Serial No. Postcard
12. ☐ Small Entity Statement(s) ☐ Statement filed in prior application. Status still proper and desired.
13. ☐ Certified Copy of Priority Document(s) (if foreign priority is claimed)
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15. If a CONTINUING APPLICATION, check appropriate box, and supply the requisite information below.

☐ Continuation ☐ Divisional ☐ Continuation-in-part (CIP) of prior application no.:  
Prior application information: Examiner: Group Art Unit:

16. Amend the specification by inserting before the first line the sentence:

- ☐ This application is a ☐ Continuation ☐ Division ☐ Continuation-in-part (CIP)  
of application Serial No. Filed on
- ☐ This application claims priority of provisional application Serial No. Filed

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TITLE OF THE INVENTION

NUCLEAR MEDICAL DIAGNOSTIC APPARATUS

CROSS-REFERENCE TO RELATED APPLICATION

5 This application is based upon and claims the benefit of priority from the prior Japanese Patent Application No. 11-063884, filed March 10, 1999; and No. 2000-57522, filed March 2, 2000, the entire contents of which are incorporated herein by reference.

BACKGROUND OF THE INVENTION

10 The present invention relates to a nuclear medical diagnostic apparatus for externally detecting gamma rays emitted from RI (Radio-Isotope) injected to a subject and generating an RI distribution in the subject on the basis of the detection data.

15 Nuclear medical diagnostic apparatuses are classified into planar image-type apparatuses which obtain an RI distribution on a projection plane and ECT-type (Emission Computed Tomography-type) apparatuses which obtain an RI distribution on a slice.

20 The ECT type nuclear medical diagnostic apparatuses include a SPECT (Single Photon Emission Computed Tomography) apparatus using single photon RI such as  $^{99m}\text{Tc}$  or  $^{111}\text{In}$ , and a PET (Positron Emission computed Tomography) apparatus using positron RI such as  $^{11}\text{C}$  or  $^{13}\text{N}$ . Recently, apparatuses serving as both SPECT  
25 apparatuses and PET apparatuses have appeared. These apparatuses in general will be called nuclear medical

diagnostic apparatuses hereinafter.

A conventional nuclear medical diagnostic apparatus has an Anger type radiation detector. As shown in FIG. 1, the Anger type radiation detector is comprised of a collimator 10, scintillator 11, lightguide 12, and plurality of PMTs (PhotoMultiplier Tubes) 13. When gamma rays come incident on the scintillator 11, fluorescence is generated at the incidence position. The fluorescence is detected by the plurality of PMTs 13. The sum of output signals from the plurality of PMTs 13 reflects the gamma ray energy. Among events wherein radiation is detected, an event derived from radio-isotope injected to the subject is selected on the basis of the total energy. The selected event is counted in association with the incidence position of the gamma rays. The incidence position of the gamma rays is calculated as, e.g., the barycentric position of energy.

Gamma rays having a high energy of 511 keV, which is derived from positron, often cause the Compton scattering in the scintillator 11. Because of the Compton scattering, energies E1 and E2 (incidence energy  $E_0 = E_1 + E_2$ ) are absorbed at two positions P1 and P2 almost simultaneously. One of the two positions P1 and P2 is the true incidence position.

Conventionally, however, the incidence position of the gamma rays is calculated as the barycentric

position of an energy which does not coincide either of  
the two positions P1 and P2 or naturally the true  
incidence position. In other words, all the events  
wherein scattering occurs in the scintillator are  
5 counted as having occurred at erroneous positions. In  
addition, conventionally, whether scattering occurs in  
the scintillator cannot be determined.

In a PET-exclusive apparatus having a BGO (bismuth  
germanium oxide) detector for performing block  
10 detection as well, when gamma rays are scattered among  
blocks of the BGO detector, the PET-exclusive apparatus  
cannot separate events that occur simultaneously to  
obtain the accurate positions of the events by  
calculation. Accordingly, a decrease in counting  
15 precision cannot be avoided.

#### BRIEF SUMMARY OF THE INVENTION

It is an object of the present invention to  
decrease, in a nuclear medical diagnostic apparatus,  
the probability of an incidence position detection  
20 error derived from scattering in a radiation detector.

The radiation detector has a plurality of  
semiconductor cells arrayed in a matrix. Each of the  
plurality of semiconductor cells detects radiation  
separately, and outputs a signal representing the  
25 energy of radiation separately. A selection circuit  
selects, among events wherein radiation is detected,  
specific events wherein radiation derived from

radio-isotope injected to a subject is detected. In the first case wherein either one of the semiconductor cells outputs a signal, the energy of the signal is compared with a predetermined energy window. In the  
5 second case wherein two or more semiconductor cells output two or more signals substantially simultaneously, the total energy of the two or more signals is compared with the predetermined energy window. A position calculation circuit calculates, in the first case, the  
10 incidence position of radiation on the basis of the position of the semiconductor cell that outputs a signal, and in the second case, the incidence position of radiation on the basis of the position of either one of the two or more semiconductor cells. A counting  
15 circuit counts the specific events in association with the calculated incidence position. The distribution of radio-isotope in the subject is obtained on the basis of this counting result.

Additional objects and advantages of the invention  
20 will be set forth in the description which follows, and in part will be obvious from the description, or may be learned by practice of the invention. The objects and advantages of the invention may be realized and  
obtained by means of the instrumentalities and  
25 combinations particularly pointed out hereinafter.

#### BRIEF DESCRIPTION OF THE SEVERAL VIEWS OF THE DRAWING

The accompanying drawings, which are incorporated

in and constitute a part of the specification,  
illustrate presently preferred embodiments of the  
invention, and together with the general description  
given above and the detailed description of the  
5 preferred embodiments given below, serve to explain the  
principles of the invention.

FIG. 1 is a sectional view of a conventional Anger  
type gamma camera;

FIG. 2 is a view showing the frequency  
10 distribution of the Compton scattering angle with  
respect to the energy of the incidence gamma rays;

FIG. 3 is a graph showing the relationship between  
an incidence energy and the energy of scattered rays at  
various scattering angles;

15 FIG. 4 is a schematic sectional view of a  
radiation detector used in a nuclear medical diagnostic  
apparatus according to an embodiment of the present  
invention;

FIG. 5 is a block diagram showing the arrangement  
20 of the nuclear medical diagnostic apparatus having the  
radiation detector shown in FIG. 4;

FIG. 6 is a view showing two positions where an  
energy caused by one scattering event in semiconductor  
cells is absorbed according to this embodiment;

25 FIG. 7 is a view showing three positions where an  
energy caused by two scattering events in the  
semiconductor cells is absorbed according to this

embodiment;

FIG. 8 is a schematic view showing the arrangement of a nuclear medical diagnostic apparatus having two opposing detectors according to this embodiment; and

5        FIG. 9 is a view for explaining a gamma ray absorption correcting method utilizing backscattered rays in the apparatus of FIG. 8.

#### DETAILED DESCRIPTION OF THE INVENTION

10        The embodiment of the present invention will be described with reference to the accompanying drawings.

First, the principle of how to reduce the probability of an incidence position detection error derived from the Compton scattering in a radiation detector will be briefly explained.

15        FIG. 2 shows the frequency distribution of the Compton scattering angle with respect to the energy of incidence gamma rays. Referring to FIG. 2, for example, when the energy of the incidence gamma rays is 511 keV ( $\alpha = 1$ ), this scattering is mostly forward scattering  
20        having a scattering angle of  $90^\circ$  or less. This tendency also applies to a case wherein the incidence gamma rays have an energy of 250 keV or more.

FIG. 3 shows the relationship between an incidence energy and the energy of scattered rays at various  
25        scattering angles ( $0^\circ$ ,  $5^\circ$ ,  $10^\circ$ ,  $20^\circ$ ,  $30^\circ$ ,  $45^\circ$ ,  $60^\circ$ ,  $90^\circ$ ,  $120^\circ$ , and  $180^\circ$ ). In FIG. 3, the axis of abscissa represents the incidence energy ( $E_0 (= E_1 + E_2)$ ), and

the axis of ordinate represents the energy ( $E_2$ ) of the Compton scattered ray. From FIG. 3, it is obvious that when the incidence energy is 511 keV, that is, when these gamma rays are generated by positron, the energy  $E_2$  of the scattered ray falls within a range:

$$170 \text{ keV } (\theta = 180^\circ) \leq E_2 < 511 \text{ keV } (\theta = 0^\circ)$$

When the scattering energy  $E_2$  falls within a range:

$$170 \text{ keV} \leq E_2 < 255 \text{ keV } (511 \text{ keV} \times 1/2)$$

then a scattering angle  $\theta$  falls within a range

$$75^\circ \leq \theta < 180^\circ$$

It is accordingly understood that 15% (painted portion in FIG. 2) of all the scattering events represents events having a scattering angle  $\theta$  which falls within the range of  $75^\circ \leq \theta < 180^\circ$ . More specifically, it is concluded that, when gamma rays having an energy of 511 keV are scattered in the radiation detector only once, 85% of its scattering energy  $E_2$  is 256 keV (1/2 of 511 keV) or more. In other words, of the two energy absorption positions, the position where less energy is absorbed is determined as the scattering position (incidence position) with a probability of 85%.

This probability varies depending on the thickness and shape of the radiation detector. Simulation such as Monte Carlo simulation is performed in which the thickness and shape of the radiation detector are initialized. Through this simulation, the detection



surface can be divided into areas having a high probability that the position with a less energy is the incidence position, and areas having a high probability that the position with a larger energy is the incidence position. Therefore, the first rule according to which the position with the less energy is selected as the incidence position, and the second rule according to which the position with the larger energy is selected as the incidence position, can also be selectively employed in units of areas.

When this determination method is employed, according to the present invention,  $1/2$  or more of the scattering events can be counted as having occurred at the true incidence positions, whereas the conventional Anger type gamma camera counts all the scattering events as having occurred at erroneously detected positions.

According to another method of reducing the probability of an incidence position detection error, when scattering occurs in a detector, i.e., when two or more semiconductor cells of one detector output signals substantially simultaneously, this event is excluded from the counting target. With this method, although the counting efficiency decreases more or less, the position detection error ratio can be suppressed to almost zero.

FIG. 4 is a schematic sectional view of a

semiconductor type radiation detector used in a nuclear medical diagnostic apparatus according to a preferable embodiment of the present invention. The radiation detector has a collimator 10, semiconductor cell array 20, and detection processing circuit 21. The semiconductor cell array 20 is formed on the rear surface of the collimator 10. The detection processing circuit 21 is formed on the rear surface of the semiconductor cell array 20. The semiconductor cell array 20 has a plurality of semiconductor cells 22 arranged in a matrix. The detection processing circuit 21 has a plurality of pre-amplifiers 23. The plurality of pre-amplifiers 23 respectively correspond to the plurality of semiconductor cells 22. The pairs of semiconductor cells 22 and pre-amplifiers 23 can detect radiation separately and output signals representing the energy of radiation separately. When the nuclear medical diagnostic apparatus is a coincidence PET apparatus, no collimator 10 is mounted on it.

The semiconductor cells 22 are made of, e.g., cadmium telluride (CdTe) or cadmium zinc telluride (CdZnTe). In place of the semiconductor cell array 20, a scintillation sensor formed by combining a scintillator (e.g., sodium iodide (NaI), LSO (Lutetium oxyorthosilicate), BGO (bismuth germanium oxide), and cesium iodide (CsI)) and a photoelectric conversion element (e.g., a photodiode) can be provided.

FIG. 5 is a block diagram showing the arrangement of the nuclear medical diagnostic apparatus having two opposing radiation detectors each shown in FIG. 4. The nuclear medical diagnostic apparatus shown in FIG. 5 according to this embodiment serves as both a single photon emission computed tomography (SPECT) apparatus and a coincidence positron emission computed tomography (PET) apparatus. The present invention can be applied to any one of a gamma camera, an SPECT apparatus, and a PET apparatus which generate an RI distribution (planar image) on a projection plane.

Two radiation detectors 50 and 51 are arranged to oppose each other through a subject. One radiation detector 50 has a semiconductor cell array 20 and detection processing circuit 21. The other radiation detector 51 also has a semiconductor cell array 30 and detection processing circuit 31.

Output signals (signals representing energies) from the detection processing circuits 21 and 31 are supplied to a signal processing circuit 40. The signal processing circuit 40 selects, among all the events wherein gamma rays are detected, a specific event (target event) wherein gamma rays derived from radio-isotope injected to the subject are detected is selected.

More specifically, in the first case, a signal is output from either one semiconductor cell 22 of each of

the radiation detectors 50 and 51. In this case, the energy of the signal is compared with a predetermined energy window. When the signal energy falls within the predetermined energy window, this event is counted as a target event in association with the incidence position or incidence path.

In the second case, two or more signals are output from two or more semiconductor cells 22 of one of the radiation detectors 50 and 51 because of the Compton scattering or the like (internal coincidence event). In this case, the energies of the two or more signals output from the radiation detector 50 or 51 substantially simultaneously are added, and their total energy is compared with the energy window. When the signal energy falls within the predetermined energy window, this event is counted as a target event in association with the incidence position or incidence path.

An internal coincidence circuit 46 calculates the time differences between the signal output from either one of the plurality of semiconductor cells 22 of one of the radiation detectors 50 and 51 and the signals output from the remaining semiconductor cells 22, and compares each time difference with a predetermined threshold. When the time difference is smaller than the predetermined threshold, the internal coincidence circuit 46 determines that this event falls under the

second case (internal coincidence event), and outputs this determination result to the signal processing circuit 40.

5 In the first case (external coincidence event), an incidence position calculating circuit 43 calculates the incidence position of the gamma rays on the basis of the position of the semiconductor cell 22 that has output a signal. More specifically, the incidence position calculating circuit 43 calculates the central  
10 position of a semiconductor cell 22 that has output a signal as the incidence position of the gamma rays.

In the second case (internal coincidence event), the incidence position calculating circuit 43 calculates the incidence position of the gamma rays on  
15 the basis of the position of either one of the semiconductor cells 22 that have output signals substantially simultaneously. More specifically, the incidence position calculating circuit 43 calculates the central position of one semiconductor cell 22,  
20 selected from the plurality of semiconductor cells 22 that has output signals according to a predetermined rule, as the incidence position of the gamma rays.

An image reconstructing circuit 41 reconstructs a tomographic image (SPECT image or PET image) on  
25 the basis of an output from the signal processing circuit 40.

When the time difference between the signals

output from the detection processing circuits 21 and 31  
in PET counting is equal to or less than a predeter-  
mined threshold, an external coincidence circuit 42  
checks whether an event wherein gamma rays are detected  
5 is a coincidence event (external coincidence event)  
wherein gamma rays derived from radio-isotope injected  
to a subject are detected. If so, the external  
coincidence circuit 42 outputs a signal representing an  
external coincidence event to the signal processing  
10 circuit 40. The signal processing circuit 40 counts  
this external coincidence event in association with an  
incidence path.

An incidence path calculating circuit 44  
calculates a straight line connecting the incidence  
15 position of one radiation detector 50 and the incidence  
position of the other radiation detector 51, both of  
which have been calculated by the incidence position  
calculating circuit 43 during PET radiography, as the  
incidence path of the gamma rays. A displaying unit 45  
20 displays a SPECT image or PET image obtained by image  
reconstruction of the image reconstructing circuit 41.

FIG. 6 shows the second case wherein the gamma  
rays are scattered in the semiconductor cells 22 of  
the radiation detector 50 or 51 and their energy is  
25 absorbed at two points P1 and P2. In this case,  
signals are respectively output from semiconductor  
cells 22 corresponding to the positions P1 and P2

almost simultaneously. The energy absorbed at the position P1 is denoted as E1, and the energy absorbed at the position P2 is denoted as E2. Either one of the two positions P1 and P2 is the true incidence position.

5 (Event Selection)

The signal processing circuit 40 first adds the energies E1 and E2 to obtain the total energy ( $E1 + E2$ ). The signal processing circuit 40 then checks whether a relationship  $E_c - W < E1 + E2 < E_c + W$  is satisfied, that is, whether the total energy falls within the predetermined energy window.  $E_c$  is the energy of gamma rays as the imaging target. When the energy of the gamma rays derived from positron is the target,  $E_c$  is set at 511 (keV). W is a value corresponding to 1/2 the window width of the predetermined energy window, and typically corresponds to about 5% to 10% of  $E_c$ .

When the above relationship is not satisfied, this event is regarded as an event other than incidence of gamma rays (random coincidence, rays scattered in the body, or the like), and is excluded from the counting target. When the above relationship is satisfied, this event is calculated as a target event in association with the incidence position and incidence path.

When the signal processing circuit 40 determines that this event falls under the second case, this event is excluded from the counting target regardless of whether it is a target event. Namely, an event falling

under the second case need not be counted. In this case, although the counting efficiency is decreased, the incidence position detection error ratio can be completely decreased to zero.

5           The energy window described above can change depending on positions. In this case, the precision of the calculated incidence position can be improved remarkably. For example, the photoelectric absorption probability of 511-keV positron nuclide of a  
10       semiconductor cell, at a portion having a thickness of about 10 mm, made of cadmium telluride (CdTe) or cadmium zinc telluride (CdZnTe) described above is about 7.4%, and its scattering probability is about 28.5%. A probability that the energy of gamma rays  
15       which have been scattered once is absorbed in the semiconductor cell array 20 is present with a proportion unnegligible when compared to the photoelectric absorption probability described above. Therefore, if the calculating method described above is  
20       employed, the same effect as the equivalent improvement of the detection sensitivity (improvement of the count) can be obtained, when compared to a case wherein all the incidence positions of the gamma rays are erroneously calculated in an Anger type gamma camera.  
25       (Position Calculation)

          In the first case, the gamma ray incidence position is calculated on the basis of the position of



one semiconductor cell 22 that has output the signal.  
More specifically, the central position of one  
semiconductor cell 22 that has output the signal is  
calculated as the gamma ray incidence position.

5           In the second case, the gamma ray incidence  
position is calculated on the basis of the position of  
either one semiconductor cell 22 among two or more  
semiconductor cells 22 that have output the signals  
substantially simultaneously. More specifically, the  
10       central position of the semiconductor cell 22, among  
the plurality of semiconductor cells 22 that have  
output signals, that has output a signal having the  
lowest energy is calculated as the gamma ray incidence  
position. According to this rule, the true incidence  
15       position can be obtained with a probability much higher  
than 50%, as described above.

          According to another rule, calculation may be  
performed in the following manner. When the plurality  
of semiconductor cells 22 that have output signals  
20       substantially simultaneously are located in the first  
area of the detection surface, the central position of  
the semiconductor cell 22, among these semiconductor  
cells 22, that has output a signal having the lowest  
energy is calculated as the gamma ray incidence  
25       position. When the plurality of semiconductor cells 22  
that have output signals substantially simultaneously  
are located in the second area of the detection surface,

the central position of the semiconductor cell 22, among these semiconductor cells, that has output a signal having the highest energy is calculated as the gamma ray incidence position.

5           During coincidence counting, when gamma rays (gamma rays derived from positron) coming incident on the semiconductor cell array 30 in the radiation detector 51 are scattered and absorbed once, the gamma ray incidence position is calculated in accordance with  
10           the same calculation scheme as that described above.

FIG. 7 shows a case wherein gamma rays are scattered at two positions. In this case, the energy is absorbed at three positions P1, P2, and P3. Namely, three semiconductor cells 22 output signals  
15           substantially simultaneously. The three signals respectively represent energies E1, E2, and E3 (keV).  
(Event Selection)

The signal processing circuit 40 adds the energies E1, E2, and E3, and compares their total energy  
20           (E1 + E2 + E3) with a predetermined energy window. The signal processing circuit 40 then checks whether the total energy (E1 + E2 + E3) satisfies a relationship:

$$E_c - W < E_1 + E_2 + E_3 < E_c + W$$

If this relationship is not satisfied, this event is  
25           excluded from the counting target. If this relationship is satisfied, this event is counted as a target event in association with the incidence position

and incidence path.

(Position Calculation)

5       The incidence position calculating circuit 43  
selects, from the three semiconductor cells 22 that  
have output signals substantially simultaneously, one  
semiconductor cell 22 that has output a signal  
representing the minimum energy among the energies E1,  
E2, and E3, and calculates the central position of the  
selected semiconductor cell 22 as the incidence  
10       position. Alternatively, the incidence position  
calculating circuit 43 calculates the middle point of  
the central positions of the two semiconductor cells 22  
that have output signals representing the two energies,  
obtained by excluding the maximum energy from the  
15       energies E1, E2, and E3, as the incidence position. To  
select which calculation scheme can be changed in  
accordance with the incidence energy.

For example, assume that the gamma rays, derived  
from positron, coming incident on the radiation  
20       detector, and scattered and absorbed first have the  
maximum energy. In this case, backscattering is  
dominant, and the two energies of the gamma rays  
absorbed after backscattering are small, so that the  
range is short in average. It is supposed that even if  
25       the two detection positions where these two energies  
are detected are averaged, a fluctuation in the  
calculated incidence position is small in average.

Assume that the gamma rays that are scattered the second time have the maximum energy. In the first scattering, forward scattering is dominant. If the detection positions of the two energies absorbed after first and last scattering are simply averaged, an incidence position more accurate in average than that obtained by weighted addition of the respective energies generated in an Anger type gamma camera can be obtained.

Assume that the gamma rays that are scattered the third time have the maximum energy. In two initial scattering cycles, forward scattering is dominant, and the range of the second scattering is long. Hence, when the detection positions of the two energies absorbed after two initial scattering cycles are simply averaged, the precision of the incidence position is largely improved.

As shown in FIG. 7, the probability that scattering occurs twice is much smaller than the probability that scattering occurs once. Yet, this can improve the precision of the calculated incidence position more than in the case using the Anger type gamma camera. In this manner, the calculation process of the incidence position as shown in FIGS. 6 and 7 can be applied to gamma rays which can cause forward scattering with a high probability (i.e., to gamma rays having a comparatively high energy).

Above explanation refers to calculation of the incidence position of gamma rays when relatively small energies are detected at two detection positions. When relatively small energies are detected at three or more detection positions, the barycenter of these detection positions may be calculated, and their barycentric position as the calculation result may be determined as the gamma ray incidence position.

FIG. 8 is a view showing the schematic arrangement of a gamma camera as a nuclear medical diagnostic apparatus having two opposing detectors (an apparatus in which radiation detectors are arranged to oppose each other through a subject) according to the embodiment of the present invention, and explains a positron imaging method using this gamma camera.

FIG. 8 is based on the following assumption. One gamma ray generated by positron Po comes incident on the radiation detector 50, is scattered once, and is then absorbed. The other gamma ray comes incident on the radiation detector 51 and is back-scattered at a scattering angle of  $\theta$ . After that, backscattered gamma rays concerning the remaining energy come incident on the radiation detector 50 entirely and are absorbed. A gamma ray incidence path is calculated on this assumption. More specifically, FIG. 8 shows a case wherein three events occur in the radiation detector 50 simultaneously, whereas one event occurs in the

radiation detector 51.

To perform coincidence counting, outputs (trigger signals) from positron generation time detection circuits (not shown) in the detection processing  
5 circuits 21 and 31 respectively formed in the two radiation detectors 50 and 51 that oppose each other through the subject P are output to the coincidence circuit 42. Based on these trigger signals, the coincidence circuit 42 checks whether energies E2, E3,  
10 and E4 of the gamma rays absorbed in the radiation detector 50 and an energy E1 of the gamma rays absorbed in the radiation detector 51 are related to the gamma rays generated by positron Po simultaneously.

If these energies are not recognized to be related  
15 to the gamma rays coming incident on the radiation detectors 50 and 51 simultaneously (if they are not recognized as coincidence counting), information on these energies should not contribute to positron imaging. If these energies are recognized to be  
20 related to the gamma rays coming incident on the radiation detectors 50 and 51 simultaneously, the incidence position calculating circuit 43 performs the following process in response to this recognition result on the basis of the energy signals and position  
25 signals output from the detection processing circuits 21 and 31.

First, assume that backscattering occurs in the

radiation detector 51 and consequently backscattering gamma rays BS come incident on the radiation detector 50, as shown in FIG. 8. A scattering angle  $\theta$  of the gamma rays BS falls within the range of  $90^\circ \leq \theta \leq 180^\circ$ , and  $90^\circ$  scattering corresponds to about 220 keV.

Accordingly, on the basis of the energy  $E_1$  absorbed in the radiation detector 51, whether a relationship  $220 < E_1 < 511 - W$  (keV) or  $E_1 < 170$  (keV) is satisfied is checked. Note that  $W$  is the window in interest, as described above.

If the relationship  $220 < E_1 < 511 - W$  (keV) or  $E_1 < 170$  (keV) is satisfied, information on the energy  $E_1$  should not contribute to imaging. If the energy  $E_1$  falls within the range of  $170 \leq E_1 \leq 220$ , the energy  $E_1$  is added with each of the energies ( $E_2$ ,  $E_3$ , and  $E_4$ ). Namely,  $E_1 + E_2$ ,  $E_1 + E_3$ , and  $E_1 + E_4$  are calculated to acquire sums  $E_1$ ,  $E_2$ , and  $E_3$ .

It is checked whether each sum satisfies a relationship  $E_1$  ( $E_2$  or  $E_3$ )  $< 511 - W$  (keV) or  $E_1$  ( $E_2$  or  $E_3$ )  $> 511 + W$  (keV). If any sum satisfies either relationship, information on these energies should not contribute to imaging.

If a sum that satisfies a relationship  $511 - W \leq E_1$  ( $E_2$ , or  $E_3$ )  $\leq 511 + W$  (keV) exists, a position ( $x_1$ ,  $y_1$ ) in the radiation detector 51 where the energy  $E_1$  is detected is determined as the incidence position of the gamma rays derived from positron. In this case, the

sum  $E1 = E1 + E2$  satisfies the relationship  $511 - W \leq E1 \leq 511 + W$  (keV).

5 The energy  $E2$  used for determination of the gamma ray incidence position in the radiation detector 51 is excluded from the energies  $E2$ ,  $E3$ , and  $E4$  detected in the radiation detector 50, and two remaining energies  $E3$  and  $E4$  are added to acquire a sum  $E4$ .

10 On the basis of the sum  $E4$ , whether  $E4$  satisfies the relationship  $E4 < 511 - W$  or  $E4 > 511 + W$  is checked. If the sum  $E4$  satisfies this relationship, on the same principle as that of the case shown in FIG. 6, the position where a lower energy, of the two energies  $E3$  and  $E4$  that are added, is detected is determined as the incidence position where the gamma rays derived from positron come incident on the radiation detector 50. Then, the incidence path of the gamma rays derived from positron is calculated on the basis of the incidence positions on the radiation detectors 50 and 51.

20 The present invention is not limited to a case wherein three incidence events occur in the radiation detector 50. When two, or four or more incidence events occur, the same method as that described above can be used.

25 FIG. 9 explains a case wherein absorption correction of gamma rays is performed by utilizing backscattered rays, without using a special gamma ray



absorption correction ray source, on the basis of the method described by using the gamma camera shown in FIG. 8. In FIG. 9, when backscattering is caused in two radiation detectors 50 and 51 opposing each other through a subject P as shown in FIG. 8, backscattered rays BS1 and BS2 come incident on the other radiation detectors 51 and 50, respectively. In this case, the energy values of these backscattered rays can be estimated with a certain fluctuation. Hence, these backscattered rays can be supposed to be the gamma ray absorption correction ray source having these energies.

More specifically, when acquiring ordinary coincidence counting PET, in addition to utilizing the backscattered rays BS1 and BS2 as described above, if the energy distributions of the backscattered rays at a certain detection position of the gamma rays at respective angles in the radiation detectors 50 and 51, and their frequencies are estimated from a certain typical patient model, gamma ray absorption correction data can be simply formed by using this estimation. By employing this method, absorption correction of the gamma rays can be performed without specially forming absorption correction data by using a gamma ray absorption correction ray source.

The method described with reference to FIGS. 8 and 9 is not limited to a case wherein the gamma camera described above, which has two opposing detectors, is

used, but can also be applied to a gamma camera having three or more radiation detectors, a PET apparatus in which a radiation detector is arranged annularly, and the like.

5           Additional advantages and modifications will readily occur to those skilled in the art. Therefore, the invention in its broader aspects is not limited to the specific details and representative embodiments shown and described herein. Accordingly, various  
10           modifications may be made without departing from the spirit or scope of the general inventive concept as defined by the appended claims and their equivalents.

WHAT IS CLAIMED IS:

1. A nuclear medical diagnostic apparatus  
comprising:

at least one radiation detector having a plurality  
5 of semiconductor cells which are arranged in a matrix,  
detect radiation separately, and output signals  
representing an energy of the radiation separately;

a selection circuit which, in order to select,  
among events wherein the radiation is detected, a  
10 specific event wherein a radiation derived from  
radio-isotope injected to a subject is detected, in a  
first case wherein either one of said semiconductor  
cells output a signal, compares an energy of the signal  
with a predetermined energy window, and in a second  
15 case wherein not less than two semiconductor cells  
output not less than two signals substantially  
simultaneously, calculates a total energy of the not  
less than two signals and compares the total energy  
with the predetermined energy window;

20 a position calculation circuit which, in the first  
case, calculates an incidence position of the radiation  
on the basis of a position of said semiconductor cell  
that has output the signal and, in the second case,  
calculates an incidence position of the radiation on  
25 the basis of a position of either one semiconductor  
cell among said not less than two semiconductor cells;

a counting circuit configured to count the

specific event in association with the calculated incidence position; and

a circuit configured to generate a distribution of radio-isotope in the subject on the basis of a counting result.

2. An apparatus according to claim 1, further comprising an internal coincidence circuit configured to determine the second case on the basis of a time difference among a plurality of signals output from said radiation detector.

3. An apparatus according to claim 1, wherein in the second case, said position calculation circuit compares the energies of the not less than two signals in order to select either one from said not less than two semiconductor cells.

4. An apparatus according to claim 1, wherein in the second case, said position calculation circuit selects, from said not less than two semiconductor cells, one that outputs a signal representing a minimum energy.

5. An apparatus according to claim 1, wherein in the second case, said position calculation circuit selects either one from said not less than two semiconductor cells on the basis of the energy of the not less than two signals.

6. An apparatus according to claim 1, wherein in the second case, said position calculation circuit

selects, from said not less than two semiconductor cells, one that outputs a signal representing a minimum energy in a first area, and one that outputs a signal representing a maximum energy in a second area.

5           7. An apparatus according to claim 1, wherein in the second case, said position calculation circuit selects one from said not less than two semiconductor cells on the basis of the energy of the not less than two signals and the positions of said not less than two  
10 semiconductor cells.

          8. An apparatus according to claim 1, further comprising a circuit configured to calculate time differences between a signal output from either one of said plurality of semiconductor cells and signals  
15 output from remaining ones of said plurality of semiconductor cells.

          9. An apparatus according to claim 1, further comprising a circuit configured to calculate time differences between a signal output from either one of said plurality of semiconductor cells and signals  
20 output from remaining ones of said plurality of semiconductor cells, and determines the second case on the basis of the time differences.

          10. An apparatus according to claim 1, wherein  
25 each of said semiconductor cells has a layer made of cadmium telluride or cadmium zinc telluride.

          11. An apparatus according to claim 1, wherein

each of said semiconductor cells has a scintillator layer and a photoelectric conversion layer.

12. A nuclear medical diagnostic apparatus comprising:

5           at least one radiation detector having a plurality of semiconductor cells which are arranged in a matrix, detect radiation separately, and output signals representing an energy of the radiation separately;

          a selection circuit which causes, among events  
10 wherein the radiation is detected, an event wherein not less than two semiconductor cells output not less than two signals substantially simultaneously, not to contribute to imaging, and selects an event derived from radio-isotope injected to a subject on the basis  
15 of the energy of the signal,

          a position calculation circuit configured to calculate an incidence position of the radiation on the basis of positions of said semiconductor cells that output the signals;

20           a counting circuit configured to count the selected event in association of the calculated incidence position; and

          a circuit configured to generate a distribution of radio-isotope in the subject on the basis of a counting  
25 result.

13. An apparatus according to claim 12, further comprising an internal incidence circuit configured to

determine the second case on the basis of a time difference among a plurality of signals output from said radiation detector.

5        14. A nuclear medical diagnostic apparatus comprising:

          at least one radiation detector having a plurality of semiconductor cells which are arranged in a matrix, detect radiation separately, and output signals representing an energy of the radiation separately;

10           a position calculation circuit which, in a first case wherein either one of said semiconductor cells outputs a signal, calculates an incidence position of the radiation on the basis of a position of said semiconductor cell that outputs the signal and, in a  
15           second case wherein not less than two semiconductor cells output not less than two signals substantially simultaneously, calculates an incidence position of the radiation on the basis of positions of said not less than two semiconductors that output the not less than  
20           two signals substantially simultaneously;

          a counting circuit configured to count an event wherein radiation derived from radio-isotope injected to a subject is detected, in association with the calculated incidence position; and

25           a circuit configured to generated a distribution of the radio-isotope in the subject on the basis of a counting result.

15. An apparatus according to claim 14, further comprising an internal coincidence circuit configured to determine the second case on the basis of a time difference among the plurality of signals output from said radiation detector.

16. An apparatus according to claim 14, wherein in the second case, said position calculation circuit calculates a barycentric position of the positions of said not less than two semiconductor cells.

17. An apparatus according to claim 14, wherein in the second case, said position calculation circuit calculates, when said two semiconductor cells output signals substantially simultaneously, an incidence position on the basis of one of the positions of said two semiconductor cells, and when not less than three semiconductor cells output signals substantially simultaneously, a barycentric position of the positions of remaining ones of said plurality of semiconductor cells obtained by excluding said semiconductor cell that has output the signal having a maximum energy.

~~18.~~ A nuclear medical diagnostic apparatus comprising:

at least one radiation detector having a plurality of semiconductor cells which are arranged in a matrix, detect radiation separately, and output signals representing an energy of the radiation separately; and a circuit configured to calculate time differences



between a signal output from either one of said plurality of semiconductor cells and signals output from remaining ones of said semiconductor cells.

19. An apparatus according to claim 18, further comprising a circuit configured to compare the time difference with a predetermined threshold.

20. A nuclear medical diagnostic apparatus comprising:

at least one radiation detector having a plurality of semiconductor cells which are arranged in a matrix, detect radiation separately, and output signals representing an energy of the radiation separately; and

a circuit which, when not less than two semiconductor cells output not less than two signals substantially simultaneously, calculates a total energy of the not less than two signals.

21. An apparatus according to claim 20, further comprising a circuit configured to compare the total energy with a predetermined energy window.

ABSTRACT OF THE DISCLOSURE

A radiation detector has a plurality of semiconductor cells arranged in a matrix. Each of the plurality of semiconductor cells detects radiation separately, and outputs a signal representing the energy of radiation separately. A selection circuit selects, among events wherein radiation is detected, specific events wherein radiation derived from radio-isotope injected to a subject is detected. In the first case wherein either one semiconductor cell outputs a signal, the energy of the signal is compared with a predetermined energy window. In the second case wherein two or more semiconductor cells output two or more signals substantially simultaneously, the total energy of the two or more signals is compared with the predetermined energy window. A position calculation circuit calculates, in the first case, the incidence position of the radiation on the basis of the position of the semiconductor cell that outputs a signal, and in the second case, the incidence position of radiation on the basis of the position of either one of the two or more semiconductor cells. A counting circuit counts the specific events in association with the calculated incidence position. The distribution of radio-isotope in the subject is obtained on the basis of this counting result.

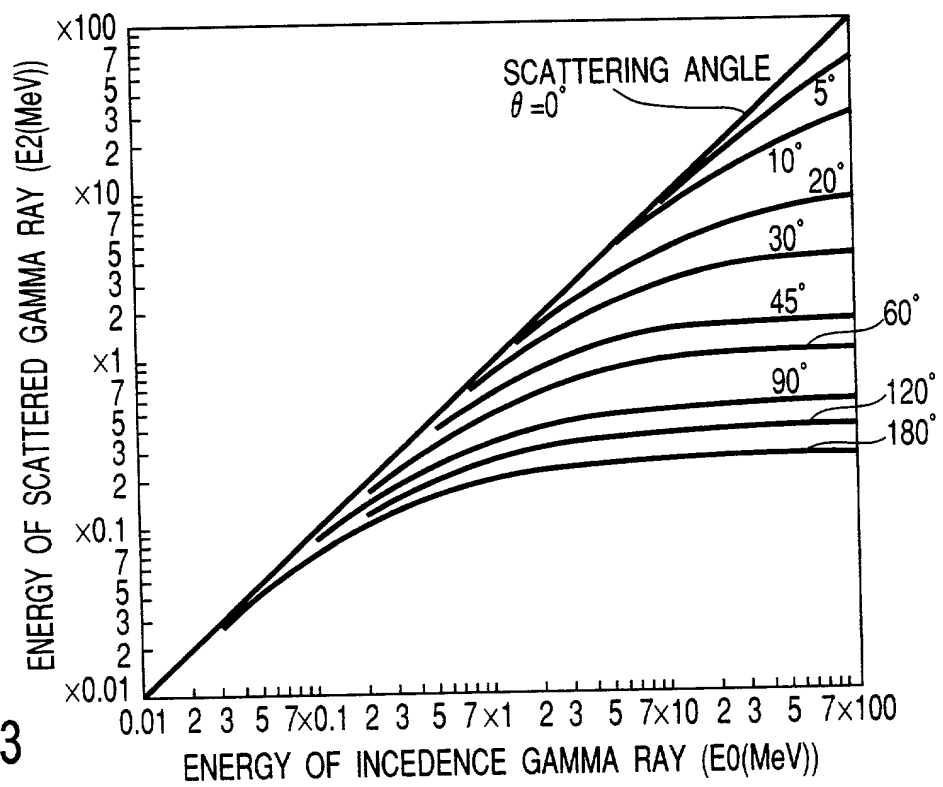
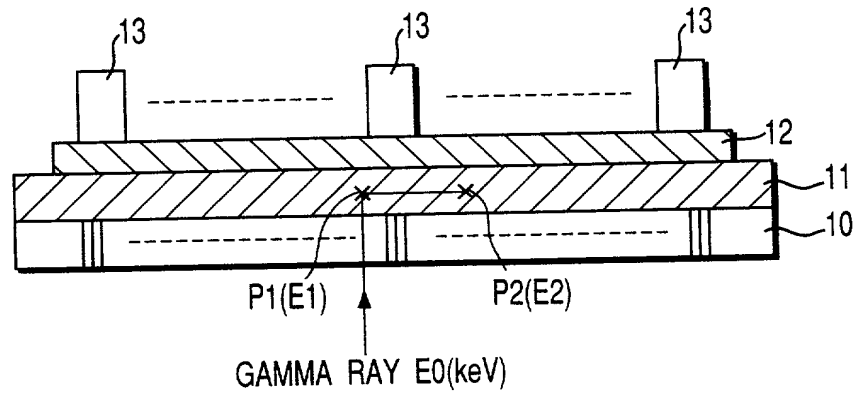
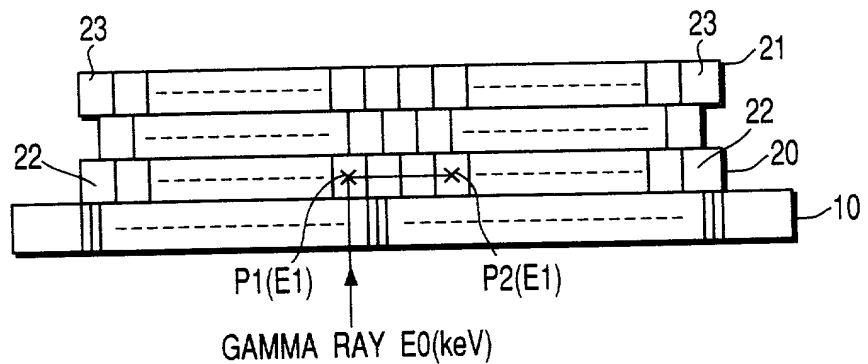
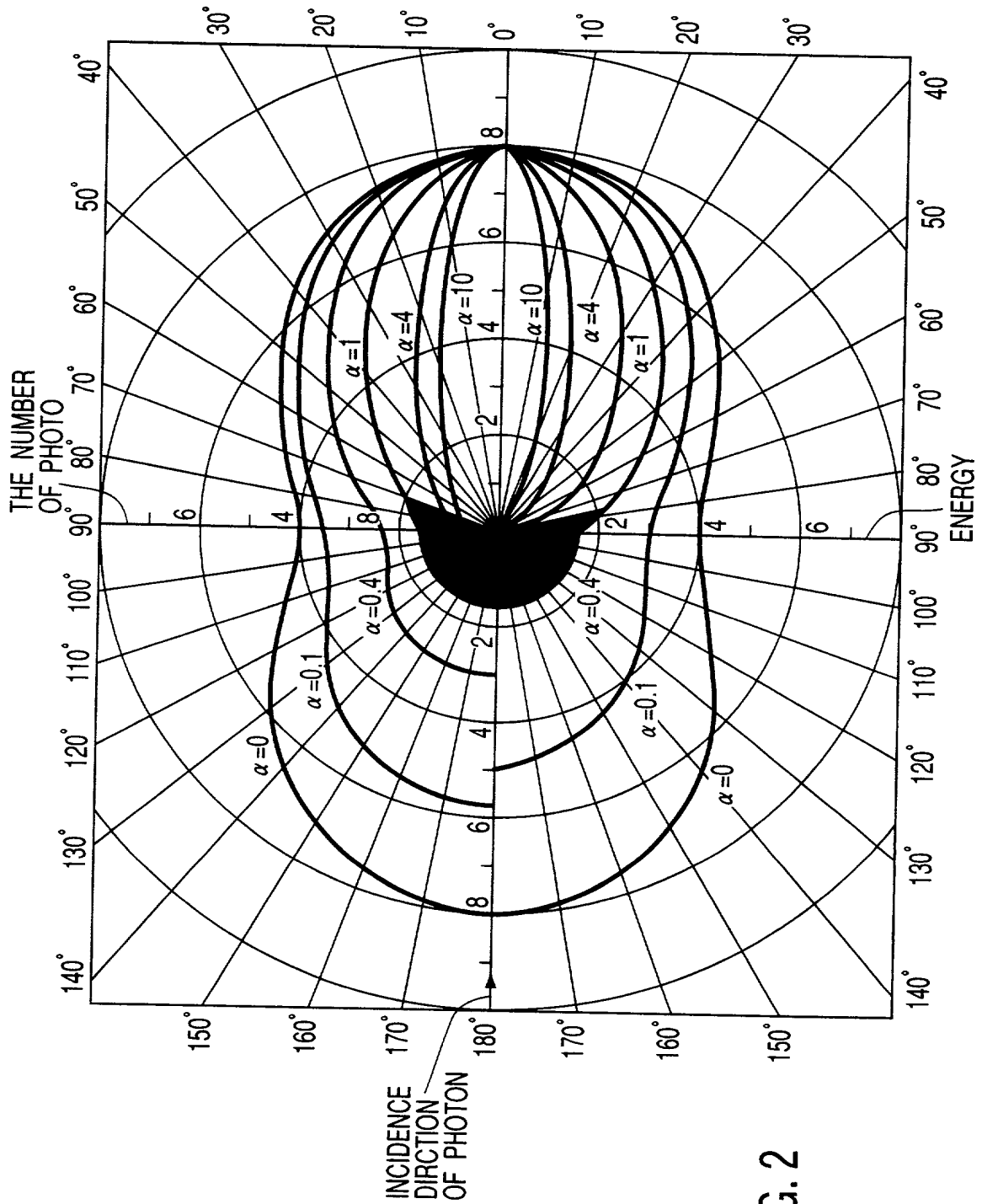
FIG. 1  
PRIOR ART

FIG. 3

FIG. 4





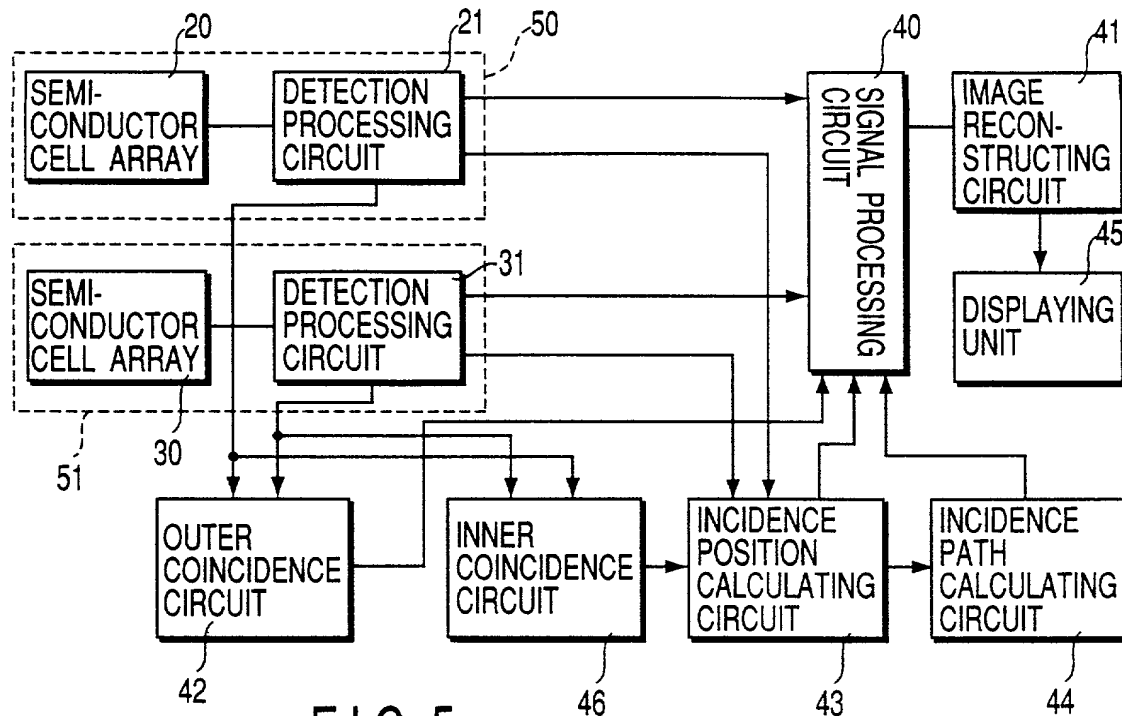


FIG. 5

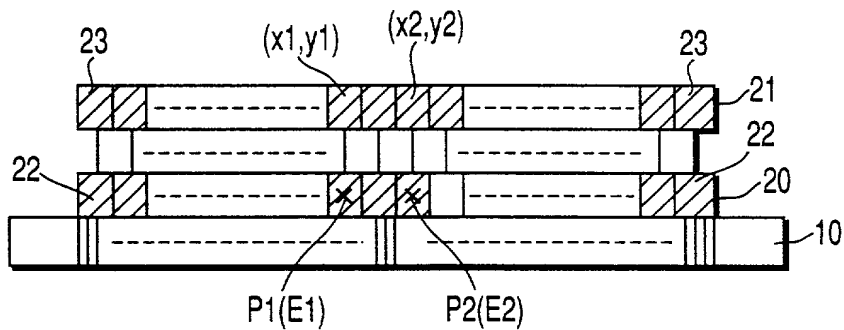


FIG. 6

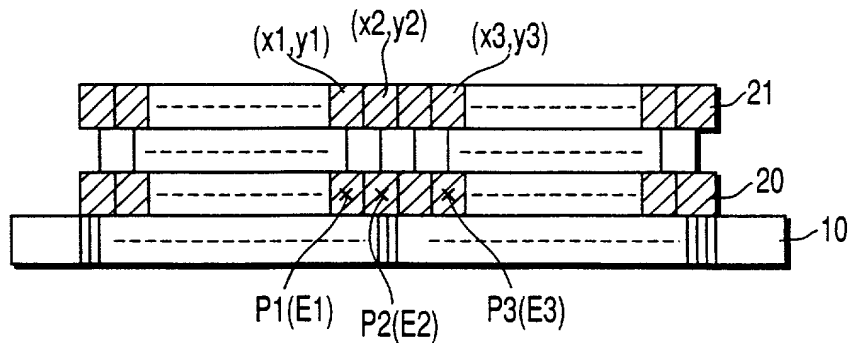
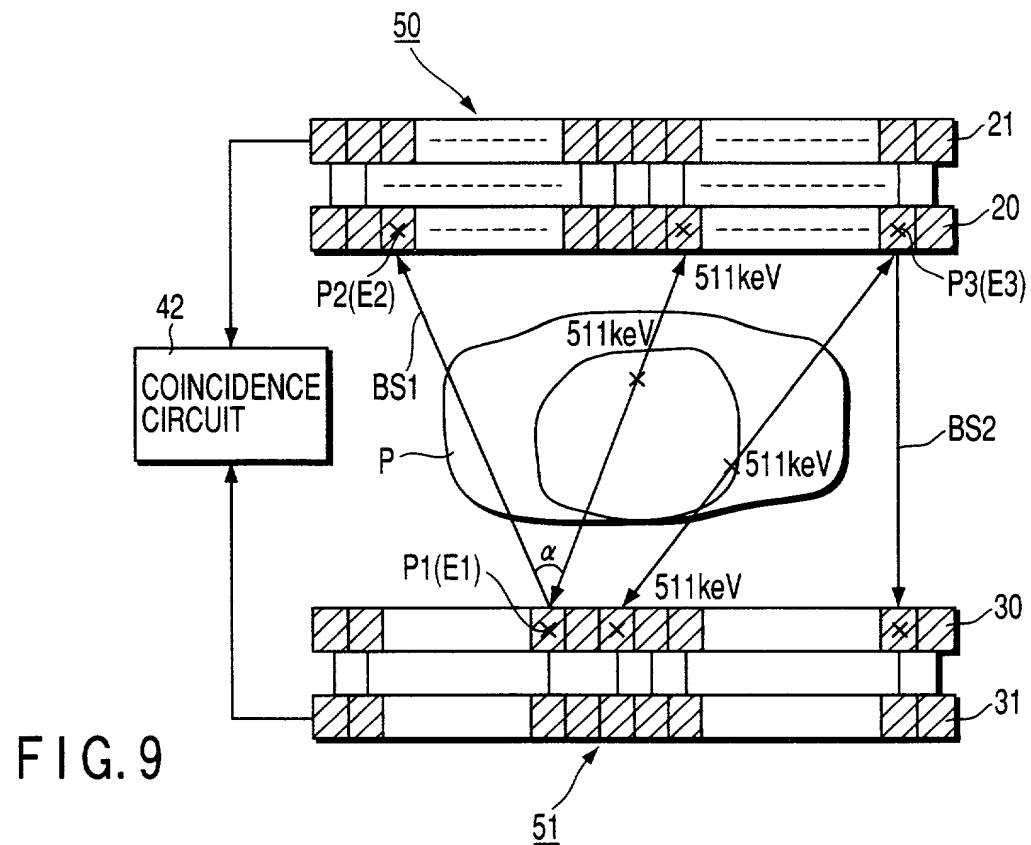
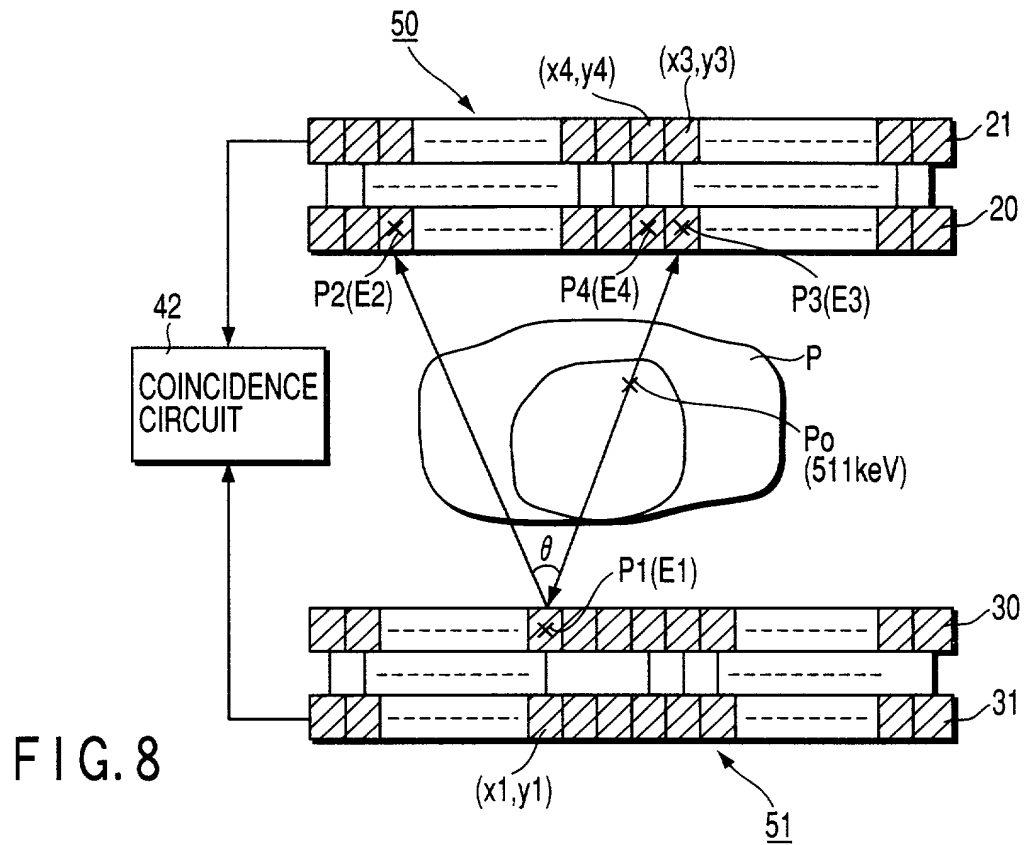


FIG. 7



## DECLARATION FOR PATENT APPLICATION

As a below named inventor, I declare:

that I verily believe myself to be the original, first and sole (if only one individual inventor is listed below) or an original, first and joint inventor (if more than one individual inventor is listed below) of the invention in

## NUCLEAR MEDICAL DIAGNOSTIC APPARATUS

the specification of which is attached hereto unless the following box is checked.

☐ was filed on \_\_\_\_\_ as United States Application  
or PCT International Application No. \_\_\_\_\_, and  
was amended on \_\_\_\_\_ (if applicable).

I hereby state that I have reviewed and understand the contents of the above identified specification, including the claims, as amended by any amendment referred to above.

I acknowledge the duty to disclose information of which is material to patentability as defined in 37 CFR 1.56.

I hereby claim foreign priority benefits under 35 U.S.C. 119(a)-(d) or 365 (b) of any foreign application(s) for patent or inventor's certificate, or 35 U.S.C. 365(a) of any PCT International application which designated at least one country other than the United States, listed below and have also identified below any foreign application for patent or inventor's certificate, or PCT International application having a filing date before that of the application on which priority is claimed:

Country	Category	Application No.	Filing Date	Priority Claim
Japan	Patent	11-063884	March 10, 1999	Yes
Japan	Patent	2000-057522	March 2, 2000	Yes

And I hereby appoint Norman F. Oblon (Reg. No. 24,618), Marvin J. Spivak (Reg. No. 24,913), C. Irvin McClelland (Reg. No. 21,124), Gregory J. Maier (Reg. No. 25,599), Arthur I. Neustadt (Reg. No. 24,854), Richard D. Kelly (Reg. No. 27,757), James D. Hamilton (Reg. No. 28,421), Eckhard H. Kuesters (Reg. No. 28,870), Robert T. Pous (Reg. No. 29,099), Charles L. Gholz (Reg. No. 26,395), Vincent J. Sunderdick (Reg. No. 29,004), William E. Beaumont (Reg. No. 30,996), Robert F. Gnuse (Reg. No. 27,295), Jean-paul Lavalleye (Reg. No. 31,451), Stephen G. Baxter (Reg. No. 32,884), Robert W. Hahl (Reg. No. 33,893), Richard L. Treanor (Reg. No. 36,379), Steven P. Weihrouch (Reg. No. 32,829), John T. Goolkasian (Reg. No. 26,142), Richard L. Chinn (Reg. No. 34,305), Steven E. Lipman (Reg. No. 30,011), Carl E. Schlier (Reg. No. 34,426), James J. Kulbaski (Reg. No. 34,648), Richard A. Neifeld (Reg. No. 35,299), J. Derek Msaon (Reg. No. 35,270), Surinder Sachar (Reg. No. 34,423), Christina M. Gadiano (Reg. No. 37,628), Jeffrey B. McIntyre (Reg. No. 36,867), Paul E. Rauch (Reg. No. 38,591), William T. Enos (Reg. No. 33,128) and Michael E. McCabe, Jr., (Reg. No. 37,182) each of whose address is Fourth Floor, 1755 Jefferson Davis Highway, Arlington, Virginia 22202, or any one of them, my attorneys with full power of substitution and revocation, to prosecute this application and to transact all business in the Patent & Trademark Office connected therewith, and request that correspondence be directed to Oblon, Spivak, McClelland, Maier & Neustadt, P.C., Fourth Floor, 1755 Jefferson Davis Highway, Arlington, Virginia 22202.

I declare further that all statements made herein of my own knowledge are true and that all statements made on information and belief are believed to be true; and further that these statements were made with the knowledge that willful false statements and the like so made are punishable by fine or imprisonment, or both, under Section 1001 of Title 18 of the United States Code and that such willful false statements may jeopardize the validity of the application or any patent issued thereon.

## DECLARATION FOR PATENT APPLICATION

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I declare further that my post office address is at c/o  
Intellectual Property Division, KABUSHIKI KAISHA TOSHIBA, 1-1 Shibaura  
1-chome, Minato-ku, Tokyo 105-8001, Japan; and  
that my citizenship and residence are as stated below next to my name:

Inventor: (Signature)DateResidenceDate: MAR. - 3. 2000Tsutomu YamakawaTsutomu YamakawaCitizen of: JapanNasu-gun, JapanDate:Citizen of: JapanDate:Citizen of: JapanDate:Citizen of: JapanDate:Citizen of: JapanDate:Citizen of: JapanDate:Citizen of: JapanDate:Citizen of: Japan